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ORIGINAL RESEARCH

ANKLE JOINT CONTROL DURING SINGLE-LEGGED BALANCE USING COMMON BALANCE TRAINING DEVICES – IMPLICATIONS FOR REHABILITATION STRATEGIES

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ABSTRACT

Background: A lateral ankle sprain is the most prevalent musculoskeletal injury in sports. Exercises that aim to improve balance are a standard part of the ankle rehabilitation process. In an optimal progression model for ankle rehabilitation and prevention of future ankle sprains, it is important to characterize different balance exercises based on level of difficulty and sensori-motor training stimulus.

Purpose: The purpose of this study was to investigate frontal-plane ankle kinematics and associated peroneal muscle activity during single-legged balance on stable surface (floor) and three commonly used balance devices (Airex®, BOSU® Ball and wobble board).

Design: Descriptive exploratory laboratory study.

Methods: Nineteen healthy subjects performed single-legged balance with eyes open on an Airex® mat, BOSU® Ball, wobble board, and floor (reference condition). Ankle kinematics were measured using reflective markers and 3-dimensional recordings and expressed as inversion-eversion range of motion variability, peak velocity of inversion and number of inversion-eversion direction changes. Peroneus longus EMG activity was averaged and normalized to maximal activity during maximum voluntary contraction (MVC), and in addition amplitude probability distribution function (APDF) between 90 and 10% was calculated as a measure of muscle activation variability.

Results: Balancing on BOSU® Ball and wobble board generally resulted in increased ankle kinematic and muscle activity variables, compared to the other surfaces. BOSU® Ball was the most challenging in terms of inversion-eversion variability while wobble board was associated with a higher number of inversion-eversion direction changes. No differences in average muscle activation level were found between these two surfaces, but the BOSU® Ball did show a more variable activation pattern in terms of APDF.

Conclusion: The results showed large kinematic variability among different balance training devices and these differences are also reflected in muscle activation variability. The two most challenging devices were BOSU® Ball and Wobble board compared to Airex® and floor. This study can serve as guidance for clinicians who wish to implement a gradual progression of ankle rehabilitation and prevention exercises by taking the related ankle kinematics and muscle activity into account.

Level of Evidence: Level 3

Keywords: Ankle sprain, EMG, kinematics, rehabilitation.

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INTRODUCTION

A lateral ankle sprain is the most prevalent musculoskeletal injury in sports.[1] This injury predominantly occurs in sports where athletes are frequently exposed to jumping and side-cutting activities.[1–3] Ankle inversion in a plantar-flexed foot position[4] or ankle internal rotation in an inverted position[5] are the most common mechanisms of ankle injury, and depending on the magnitude of the inverting moment, it may cause damage to the mechanoreceptors in the lateral ligaments and capsule.[6] Severe injuries are often associated with the presence of residual pain, giving way sensations and neuromuscular and mechanical deficits, potentially leading to chronic ankle instability (CAI).[6–8] CAI is a multifactorial condition especially associated with neuromuscular components, which often are referred to as sensori-motor deficits.[6]

An important part of sensori-motor control of the ankle is muscle activity around the ankle joint, which contributes to ankle stability.[9] Konradsen et al[6] advocated that the peroneal muscles play an important role in ankle injury protection, because they are the primary evertors of the foot and ankle complex, whereby they are able to resist inverting moments potentially leading to injury. Muscle function deficits that have been reported for the peroneal muscles after ankle injury include reduced muscle activation (electromyographic amplitude) during gait[10,11] and jumping tasks,[12] reduced evertor muscle strength,[13] and increased muscle reaction times to simulated sprains.[14] In people with CAI, these deficits are likely related to loss of sensori-motor function of the ankle, due to mechanical damage of mechanoreceptors within the ligaments and musculature and altered mechanical properties of the ligaments after the ankle trauma.[15] Rehabilitation following an ankle sprain or recurrent sprains (CAI) is therefore recommended,[16] and must target the restoration and enhancement of proprioceptive and neuromuscular abilities and strengthening of the muscles.[17,18] Sensori-motor training seems to be an important rehabilitation modality in order to improve sensori-motor function of the ankle joint and, ultimately, reduce the risk of future sprains.[19,20] Freeman was the first to propose that sensori-motor training could decrease sensori-motor deficits at the ankle by re-educating the normal mechanoreceptor pathways in the sensori-motor system.[15] Since then numerous authors have acknowledged sensori-motor training as an effective tool for minimizing the risk of recurrent lateral ankle sprains.[21–25]

When conducting sensori-motor training, the use of exercises on balance devices are a standard part of the ankle joint rehabilitation process,[17,26,27] because they enable exercise progression. When an exercise is performed on an unstable surface, a number of authors have reported increased ankle muscle activity (EMG).[28,29] In clinical practice, many different balance devices are used, such as wobble boards, soft mats, tilt boards, and BOSU® Balls.[26,30–32] As a clinician is it important to distinguish between the different devices concerning pertubation potential and intensity. But exact knowledge on how the intensity influences ankle kinematics and muscle activity is still lacking.[26]

In optimal ankle rehabilitation, secondary prevention also needs to be included in later stages to reduce the high risk of recurrence and minimize the risk of CAI.[21–24] It is therefore important to evaluate different exercises and devices based on their level of difficulty and sensori-motor training stimulus in order to optimize rehabilitation, through specific exercises and progression models, which eventually may improve the quality of care and success of return to play, potentially reducing recurrence and CAI. A logical and simple approach to this is to analyze ankle kinematics in the frontal plane, as this plane most often is implicated in the inversion injury mechanism, during execution of balance exercises and to quantify the associated muscle activation variability. So far no studies have included both ankle kinematics and muscle activity and thereby no recommendations have been made for the progression using unstable surfaces in ankle injury rehabilitation and prevention. Both variables may be important in injury rehabilitation and prevention, to choose optimal progression from one balance exercise to another.

The purpose of this study was to investigate frontal-plane ankle kinematics and associated peroneal muscle activity during single-legged balance on stable surface (floor) and three commonly used balance devices (Airex®, BOSU® Ball and wobble board).

METHODS

Subjects

Nineteen healthy subjects, 10 male and 9 female with age, body mass, and height at 28.8 ± 2.3 (range 20-31 years), 71.9 ± 11.5 (range 55.4-93.4 kg), and 177.2 ± 11.3 (range 158.3-195.5 cm), respectively, volunteered to participate in the study and were all included by convenience sampling. All subjects were active in sports, corresponding to 5.2 ± 3.0 hours per week (range 1-13.5 hours). The study included: men or women between 20 and 35 years of age who were active in sports which required frequent jumping or side-cutting movements such as soccer, handball, basketball, and Crossfit. The exclusion criteria included: any history of a lower extremity injury or any structured rehabilitation or self-directed sensori-motor training in the preceding six months, not including strength training in general. All subjects were told not to perform strength training 48 hours prior to testing, in order to avoid delayed onset muscle sourness (DOMS) during testing. Other cardiovascular activities such as running, swimming or biking were permitted. None of the subjects reported a history of neurological or vestibular impairments. According to Danish law, the local ethics committee did not need to perform a full ethics review, because the exercises were all commonly used in standard training programs and due to the non-invasive character of the study. All subjects gave their informed consent, according to the Helsinki Declaration, before participation in the study.

Test protocol

The test session included evaluation of single-legged shod balance on four different surfaces – the floor, an Airex® mat, the convex side of a BOSU® Ball, and a multi-directional wobble board (Figure 1), with a maximal tilt angle of 21°. The wobble board, Airex® and BOSU® Ball were all included as they are commonly used for injury prevention and rehabilitation of lower extremity injuries.[31-36] Furthermore, exercises on wobble board have well-documented effect on prevention of recurrent ankle sprains [21-25] and "giving way" episodes.[37] Initially, anthropometric data were collected followed by a screening of limb dominance using a performed kick-test.[38] The limb used for kicking a ball was defined as dominant and subsequently used for measurements during the session. All subjects were instructed verbally and permitted three short practice trials in each condition before completing three, valid, 15-second duration trials, which were recorded. The test protocol ensured that exercises were performed in randomized order with 30-second breaks between trials to avoid muscle-fatigue. Randomizing the order of exercises was done manually by writing down all exercises on different papers, which were folded and blinded to a drawer, who finally draw one exercise from another from a bowl for each of the subjects. During all trials subjects had eyes open and the contralateral knee was maintained flexed in a 70-90 degree angle. A trial was discarded if the subject could not keep the balance for the 15 seconds or required any correction such as re-adjusting their position by moving the foot or touching the floor/balance device with the opposite foot. During all the test sessions this only happened three times across all subjects.

DATA ACQUISITION

Kinematic data

An eight camera Vicon 612 Vcam motion capture system (Oxford Metrics Ltd., Oxford, UK) was used to track the three-dimensional trajectories of reflective markers placed on the foot, ankle and shank of the subjects. Two markers attached to the surface of the subjects' shoe corresponding to the head of first and fifth metatarsal were used to calculate the frontal plane movements of the foot, and as such be used as an estimate of the variation and amplitude of inversion and eversion movements of the ankle. Recordings were synchronized to Vicon 612 Workstation and marker positions were sampled at a frequency of 100 Hz. Prior to test protocol, a static capture during quiet standing in the anatomical position was recorded to permit the calculation of offset values for the two markers. Subsequently, differences in vertical displacement between the first metatarsal head marker (MH1) and fifth metatarsal head marker (MH5) were calculated to describe the kinematic variability of foot inversion and eversion during exercises. This means that MH5-MH1 > 0 indicates an everted position, and MH5-MH1 < 0 indicates an inverted position, and therefore increasing values indicate a movement towards eversion, and vice versa (Fig. 2).

Figure 1. *Shows a subject performing single legged balance on the different surfaces. A: Floor, B: Airex®, C: Wobble board and D: BOSU® Ball.*

EMG data

EMG signals were recorded using rectangular (20 mm x 30 mm) bipolar surface electrodes (DE-2.1, Delsys, Boston, MA, USA). The electrodes were applied to peroneus longus according to the guidelines of Perotto et al.[39] Skin surfaces were shaved, abraded and cleaned with alcohol to improve the conductivity of the EMG signals.[39,40] All EMG signals were collected in a box (Myomonitor IV, Delsys, Boston, MA, USA) attached at the back of the subject. Here, data were amplified and band-pass filtered between 15-450 Hz, sampled at 1000 Hz, and wirelessly transmitted to a computer with a fixed delay of 200 ms and thereby converted from analogue-to-digital via a 64-bit A/D converter in the Vicon 612 Workstation.

Prior to the experimental procedures, overall maximal EMG value of the peroneus longus muscle (EMGmax) found during two trials of maximum voluntary contractions (MVC), was used as a reference value for normalization of peroneus longus activity during exercises. The MVC was assessed and measured for five seconds with the subject in supine position performing a maximal ankle eversion against

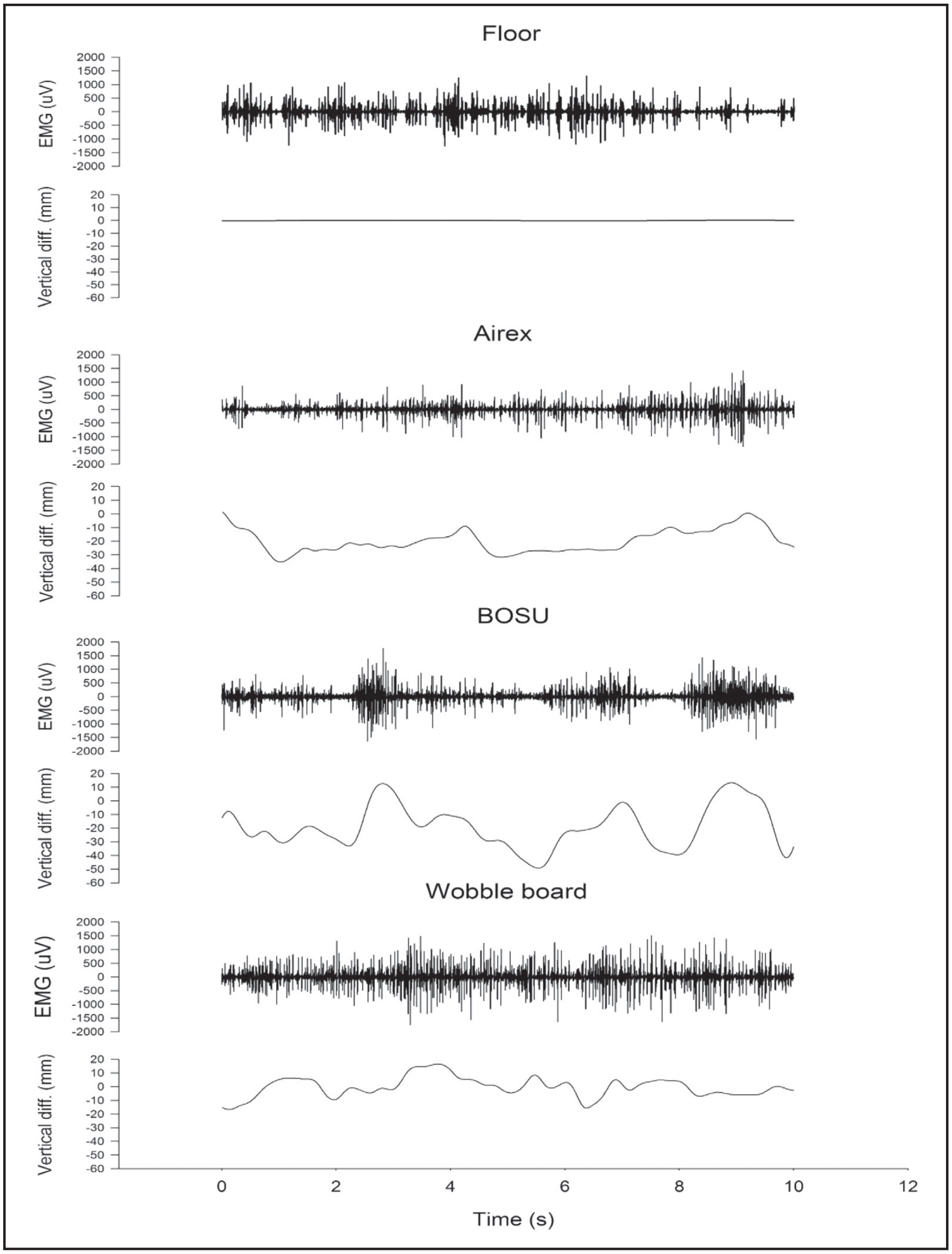

Figure 2. *Shows kinematic data and muscle activity during a single subject trial on the four different surfaces. The two measurements are synchronized during the 10 sec trial. > 0 indicates an everted position, and MH5-MH1 < 0 indicates an inverted position, and therefore increasing values indicate a movement towards eversion, and vice versa.*

manual resistance according to Kendall et al.[41] During each trial, subjects were instructed to contract "as forcefully as possible with a gradual increase in force," and strong standardized verbal encouragement was provided during the contraction. Subjects were given a 30 second break between trials.

Data reduction

For all trials, the first two and a half and last two and a half second of the 15 second recordings were discarded to ensure no postural adjustments in the beginning were captured, and to avoid muscle-fatigue in the end.[42] Previously, Harput et al.[43]

have shown good to excellent reliability using this procedure in balance tests. All data were recorded, synchronously in Vicon 612 Workstation and subsequently processed offline using a custom-written matlab script (MATLAB, version 7.2).

Kinematic data

For each trial, the difference in height of the two reflective markers attached over the first and fifth metatarsal head were used to express the following frontal-plane kinematic variables: inversion-eversion variability was calculated as the standard deviation of the difference in marker height. The inversion-eversion variability represents the variation of the movements and could be an overall expression of instability. Inversion peak velocity was calculated as the mean peak velocity of all inversion movements occurring during the 10-second trial. The velocity represents the intensity severity of the movements and a large velocity indicates a forceful stimulus. The inversion-eversion direction changes were calculated as the total number of changes in direction in the frontal plane. This is an expression of the frequency of stimuli, affecting the sensori-motor system.

EMG data

For each trial, muscle activity was calculated as the average, normalized EMG amplitude to indicate average muscle activation, and muscle activation variability was calculated as the difference between the 90 and 10 % probability level of muscle amplitude probability distribution functions (APDF).[44] A high value for this difference reflects high muscle activation variability over time. The raw EMG-signals of both static MVC-trials and balance trials were calculated as RMS-values (root-mean square) using gliding windows of 100 ms, with 99 ms overlaps.[40]

Statistical procedures

Because this study used an exploratory design, no a priori sample size estimation was performed. In this study design, the subjects served as their own controls between the different test situations. In this way, the possible variability in kinematic data due to inaccurate marker placements observed in other intervention studies would not influence the results, as the markers remained in the same placement during all test situations. Data analysis was performed using SPSS 12.0 for Windows XP. Distributions of variables are presented as mean ± one standard deviation (SD). All data were statistically examined for normality of distribution using the Kolmogorov-Smirnov test for normality. Data were normally distributed and thus parametric statistics were applied. A one-way Repeated measures ANOVA of subjects during the four different conditions was applied, sphericity assessed, and Bonferroni correction was made according to the number of comparisons that were made and a level of $p < 0.05$ was chosen to indicate statistical significance.

RESULTS

Frontal-plane kinematics

Ankle inversion-eversion variability was significantly different between groups ($p < 0.001$). Specifically, balancing on BOSU® Ball was more challenging (17 mm ± 5 mm) compared to floor (1 mm ± 1, $p < 0.001$), Airex® (9 mm ± 2, $p < 0.001$), and Wobble board (12 mm ± 6, $p < 0.001$). Also, ankle inversion-eversion variability showed significant differences when balancing on Wobble board and Airex® compared to balancing on floor ($p < 0.001$). No other differences in ankle inversion-eversion variability were observed (Table 1).

Analyzing the ankle inversion peak velocity of each exercise, showed significant different peak values when balancing on BOSU® Ball (83 mm/s ± 28) and Wobble board (67 mm/s ± 38) compared to floor (2 mm/s ± 5, $p < 0.001$) and Airex® (39 mm/s ± 16, $p < 0.01$). Balancing on Airex also yielded significant different inversion peak velocity compared to just balancing on floor ($p < 0.001$). No other differences in inversion peak velocity were observed (Table 1).

The number of ankle inversion-eversion direction changes showed significant differences when balancing on Wobble board (37 ± 9) compared to floor (4 ± 3, $p < 0.001$), Airex® (19 ± 6, $p = 0.000$) and BOSU® Ball (26 ± 7, $p = 0.000$). Also, the number of ankle inversion-eversion direction changes when balancing on BOSU® Ball showed significantly differences compared to floor ($p < 0.001$) and Airex® (0.000), and furthermore balancing on Airex® yielded significant different number of direction changes compared to balancing on floor ($p < 0.001$) (Table 1).

Table 1. *Ankle kinematics reported as mean and (standard deviation) during single-legged balance in four different conditions (n = 19)*

Exercise	Ankle IV-EV variability (mm±SD)	Ankle IV peak velocity (mm/s±SD)	Ankle IV-EV direction changes (number±SD)
Floor	1.0. ± 1.4	2.3 ± 5.5	4.2 ± 2.5
Airex®	9.5 ± 2.0#	38 ± 16#	18.9 ± 6.6#
Wobble board	11.8 ± 6.0#	67 ± 29*	37.2 ± 9.4†
BOSU® Ball	16.9 ± 5.6†	83 ± 38‡	26.5 ± 7.3‡

IV-EV = Inversion-Eversion † Indicates significantly larger than Wobble board, Airex and floor (p<0.001), ‡ Indicates significantly larger than Airex and floor (p<0.001), * Indicates significantly larger than Airex (p<0.001) and floor (p<0.001), # Indicates significantly larger than floor (p<0.001).

Peroneal muscle activity

Peroneal muscle relative activity was significantly higher when balancing on BOSU® Ball (32 % EMGmax ± 12 %) and Wobble board (36 % ± 14) compared to floor (21 % ± 8, $p < 0.001$) and Airex® (22% ± 8, $p < 0.001$). No other differences were observed in peroneal muscle relative activity levels (Table 2).

Peroneal muscle activation variability (APDF) was significantly greater balancing on BOSU® Ball (41 % ± 16) compared to floor (31 ± 13, $p < 0.002$), Airex® (32 ± 13, $p < 0.01$) and Wobble board (33 ± 13, $p < 0.001$). No other differences among the exercises were observed (Table 2).

DISCUSSION

As a clinician involved with ankle joint rehabilitation it is important to distinguish between balance exercises based on their influence on ankle kinematics and muscle activity. The present study investigates ankle inversion-eversion kinematics and associated peroneal muscle activity during single-legged balance with eyes open in four different conditions. The main finding of the study was that balance exercises performed on an unstable surface dramatically increased ankle kinematics and subsequently the muscle activity variables compared to standing on a stable surface.

Other authors have investigated the effect of various balance devices on balance and neuromuscular

Table 2. *Normalized muscle activity (% of EMGmax) reported as mean and (standard deviation) during single-legged balance in four different conditions (n = 19)*

Exercise	EMG average muscle activation (%±SD)	EMG muscle activation variability (APDF±SD)
Floor	20.7 ± 8.5	30.6 ± 13.7
Airex®	21.9 ± 8.3	31.5 ± 12.8
Wobble board	36.1 ± 14.8‡	33.2 ± 13.8
BOSU® Ball	32.4 ± 12.6‡	41.3 ± 16.9†

ADPF: Amplitude Probability Distribution Function, † Indicates significantly larger than Wobble board, Airex and floor (p<0,009), ‡ Indicates significantly larger than Airex and floor (p<0.001).

control. Stanek et al [32] measured centre-of-pressure (COP) and average sway velocity during balance exercises on four different balance devices. Within the four tested devices, the BOSU® Ball seemed to be the most challenging one in terms of both COP and average sway velocity, although they, unlike the present study, investigated the BOSU® Ball with subjects standing on the convex side of the device. However, the latter study had some limitations, as the actual impact of the devices on ankle stability or neuromuscular response was not investigated, and furthermore a wobble board was not investigated although this is a device known to have an impact on ankle injury recurrence and CAI.[21–24] When comparing the different training devices in this study, the BOSU® Ball induced almost twice the amount of inversion-eversion variability and amplitude of perturbation compared to the least unstable training device i.e. the Airex®. The BOSU® Ball also significantly surpassed the wobble board in severity of perturbation amplitude, but in terms of inversion-eversion directional changes, the wobble board was superior to all the other devices. It is likely that the differences occurred due to the different configuration of the BOSU® Ball and wobble board which results in a smaller base of support than the more densely constructed Airex® and flat floor. For that reason, the current results support that sensori-motor training can be progressed in difficulty by systematically reducing the base of support.[26]

With regard to muscle activity, the results of the current study also show greater average muscle activation level on the surfaces with the highest angular excursion during the balance exercise i.e. the BOSU® Ball and the wobble board, and lower activation levels on surface with less stability demands. These results thereby support similar findings of increased peroneal muscle activity when an exercise is performed on a unstable surface.[28,29,45] However, the current results did not demonstrate any differences in mean peroneal muscle activity between the BOSU® Ball and wobble board, thereby supporting the findings of Harput et al. [43] This is in contrast to the study by De Ridder et al [45] who showed higher peroneal activity when balancing on a BOSU® Ball compared to a wobble board. One explanation for these differences could be that the latter study used a single-axis compared to the multi-directional wobble board used in the present study. The multi-directional wobble board seems to challenge the peroneal muscles more and therefore may explain the equal activity levels found in the present study.

Despite no differences in mean peroneal activity between balancing on wobble board or BOSU® Ball, the latter did show larger and more rapid inversion-eversion direction changes. This may demand more rapid and increased muscular activation, which was expressed in a more variable activation pattern of the peroneal muscles (greater APDF values) when balancing on the BOSU® Ball. These functional differences between two commonly used balance devices have not been reported previously.

The presence of CAI is still high and ankle sprains continue to pose a significant burden for the athletes as well as the society as a whole. The results of the present study could provide valuable information to the clinician in terms of optimizing the progression of rehabilitation programs towards return to play for the athlete. Potential future studies may investigate if the increased kinematic perturbations and greater variation of peroneal muscle activity when balancing on the BOSU® Ball compared to just training on the wobble board result in greater benefit from training on this device. This has not yet been investigated.

According to existing literature, it is evident, that rehabilitation should must include balance exercises on a wobble board to some extent.[21–25,47] The current results may offer an explanation of these positive results by demonstrating high average peroneal muscle activity when exercising on the wobble board, although the exact neuromuscular link between increased ankle joint muscular activity, when using this surface, and muscular recruitment during e.g. sports activity is not yet known. However, has been suggested that sensori-motor training induces positive adaptations in the sensori-systems assisting postural control, including the vestibular, the visual, and the somato-sensori system, as well as the motor-system controlling motor output.[30] The underlying neural adaptations have been shown to occur at different sites within the central nervous system. Sensori-motor training which increases the postural demands, seems to increase subcortical structures, while it reduces spinal reflex excitability and cortical involvement.[30]

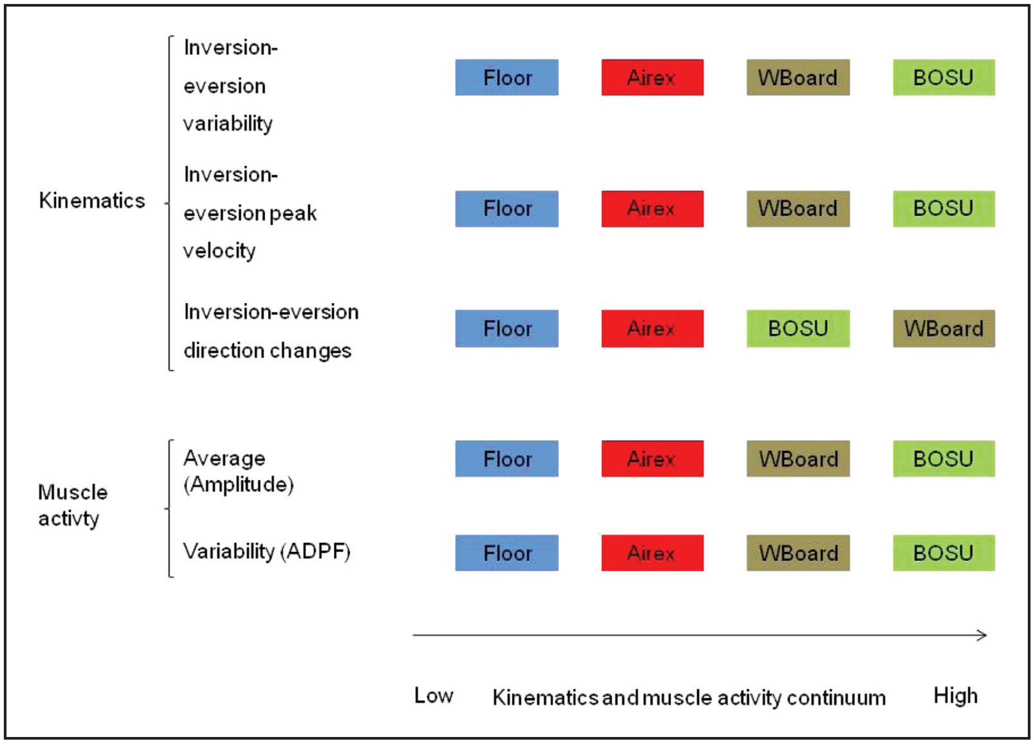

Figure 3. *Shows a summary box for both ankle kinematic data and muscle activity. The surfaces go from low to high in order to the results of the study.*

So it may be argued, that the frequent inversion-eversion directional changes and associated peroneal muscular activity monitored by proprioceptors and the protective musculature will induce an increased activation of the afferent pathways and assist in generation of a coordinated motor response and thereby establish improved stability that could be utilized in situations potentially leading to re-injury. This is corroborated by a study by Clark and Burden [48], who investigated the biomechanical effects of balance training and found faster reaction times during a perturbation as a result of a neuromuscular training programme performed utilizing a wobble board.[48]

When designing a rehabilitation program, it is necessary to challenge the sensori-motor system, normally starting with static balance exercise phase progressing to more dynamic balance exercises and then to a more functional phase facilitating more muscle pre-activity through sports specific movements.[16] The results of this study provide a guide for increasing the level of perturbation stimulus with an associated change in peroneal muscle activity during rehabilitation of ankle injuries.

The different examined balance devices may thus be ranked based on their mechanical stability demands and the resulting levels of muscle activation (Figure 3). This may be of assistance when clinicians are planning appropriate rehabilitation programs for persons with ankle joint injury. When the patient has established the ability to progress from bilateral stance to painfree single-legged balance, the results of this study indicate that rehabilitation should start with single-legged balancing on the floor and then progress to the Airex, which did not show higher levels of muscular activity but higher inversion-eversion kinematic changes in all three measurements. When the patient has established good balance on floor and subsequently the Airex®, the clinician can determine if the patient is ready to progress. The next steps would be to the wobble board and finally to the BOSU® Ball, as the present results indicated that the BOSU® Ball is more challenging than the wobble board due to the fewer but larger and more rapid direction changes in ankle inversion-eversion kinematics, resulting in a more variable activation pattern of the peroneal muscle.

The application of the results from the current study has limitations, as the study did not include all available balance devices. It is possible that other devices would have been more challenging in terms of kinematics and muscle activity. Furthermore the study was conducted using healthy subjects. However, de Ridder et al [45] did not find any differences in muscle activity (EMG) between healthy subjects and subjects with CAI, and it may be speculated that healthy subjects would have less biological variation and the results from each device therefore better express the degree of perturbation as a results of the mechanical properties of the device rather than individual differences in subject balance abilities due to pathology and it may also be argued that the basic physiological characterization of the exercises performed in the present study is best performed initially in healthy subjects where biological variation is less than in subjects with pathology.

The protocol may vary from some rehabilitation situations, as the subjects in the current study performed the exercises wearing shoes. This may add a degree of additional stability compared to bare-footed situation, but on the other hand better reflect the injury situation in actual sport. In the present study, the authors focused on the peroneus longus muscle as the most important muscle to prevent inversion trauma, and we used markers on the 1st and 5th metatarsal heads to estimate the amount of inversion-eversion. We recognize that degree of plantarflexion may influence the magnitude of estimated inversion-eversion, and that other muscles may be important for stability, especially the tibialis anterior due to its dorsiflexion function placing the ankle joint in a more closed-packed position and thereby reducing the risk of injury. However, since no differences in plantar flexion across devices were evident this may not influence the outcome significantly.

CONCLUSION

The results of the current study show large kinematic variability among commonly utilized balance training devices and that these differences are also reflected in muscle activation variability of the peroneal muscles. The two most challenging balance devices i.e. the BOSU® Ball and the wobble board, show a kinematic perturbation almost twice that achieved when using the Airex® and 8-10 times that of achieved when standing on the floor. The results of this study can serve as guidance for clinicians who wish to implement a gradual progression of ankle rehabilitation and prevention exercises by taking the related ankle kinematics and muscle activity into account.

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